THE EFFECTS OF SHOE CUSHIONING ON IMPACT FORCE DURING RUNNING

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INTRODUCTION

During heel-toe running in humans, the vertical component of the ground reaction force ($F_z$) (Cavanagh and LaFortune, 1980; Munroe et al., 1987) has characteristic “passive” and “active” peaks. The “passive” or “impact” peak ($F_{z1}$) is observed within 50 ms of initial ground contact and is associated with the impact between the body and the ground. The magnitude of $F_{z1}$ has been used as an index of impact shock, particularly in studies of running shoe cushioning. Counterintuitively, $F_{z1}$ is not reduced by softer shoe cushioning (e.g. Clarke et al., 1983; Nigg and Bahlsen, 1988; Snel et al., 1985) and may even be higher in more “cushioned” shoes (e.g. Nigg et al., 1987).

The absence of a cushioning effect on $F_{z1}$ is inconsistent with other observations. In-vitro impact test results show reductions in impact shock that are consistent with mechanical theory. Different cushioning systems can also be distinguished by in-vivo in-shoe pressure measurements (Hennig & Milani, 2000), human pendulum impacts (LaFortune et al., 1996) and subjective perception (Milani et al., 1997). There is also reason to question the validity of $F_{z1}$ as a measure of lower extremity impact. The ground reaction force reflects the vector sum of the accelerations of all the body’s segments. Kinematic analysis has shown that the impact peak has its origin in lower extremity accelerations, but that its magnitude is determined by acceleration of the rest of the body (Bobbert et al., 1991). Lower extremity acceleration itself is a combination of low frequency “active” components and higher frequency “impact” components (Shorten and Winslow, 1993). Consequently, the magnitude of $F_{z1}$ may not accurately quantify impact magnitude and the presumption that $F_{z1}$ is related to lower extremity shock may not be a robust one. Therefore, the purpose of this study was to reevaluate the nature of the impact component of $F_z$, and how it is affected by running shoe cushioning.

TWO COMPONENT MODEL OF Fz DURING RUNNING

While spectral analysis reveals high and low frequency components in $F_z$ signals, frequency domain methods (e.g. harmonic analysis, FFT, filter banks) cannot accurately quantify non-stationary pulses. Therefore, explicit modeling of $F_z$ is preferable. The conceptual division of $F_z$ into “impact” and “active” components suggests the simplest form such a model can take – a leg spring or low frequency component ($F_{LF}$) and an impact or high frequency component ($F_{HF}$), with each modeled by the impact of a mass-spring (Figure 1). The reaction force $F(t)$ during impact of a mass-spring impact with mass $m$, natural frequency $\omega_n$ and impact velocity $v_0$ is

$$F(t) / mg = (v_0 \omega_n / g) \sin (\omega_n t) + 1 - \cos (\omega_n t)$$

METHODS

Fifteen male subjects (73.3 kg ± 9.0 sd) gave their informed consent and ran at 4.0 ± 0.2 m s⁻¹ along a 40 m rubber-surfaced laboratory runway. Subjects completed four acceptable trials under each of four, randomly presented footwear conditions (Table 1a). Three otherwise identical shoe conditions (CS,CM,CH) had EVA midsoles of different hardnesses, approximating the 5th, 50th and 95th percentiles of available heel cushioning properties. The control condition (C0) was a “minimally cushioned” shoe with elastic sock upper, rubber outsole and thin foam insole but no midsole. Ground reaction forces from a single step of each trial were sampled at 1200 s⁻¹ using a runway-mounted Kistler force plate with an $F_z$ natural frequency of 450 Hz. $F_z$ data from each step were normalized to bodyweight and contact time, $T_C$. The best-fitting, two-component model of $F_z$ from each trial was determined using least squares minimization and an optimized search strategy.

RESULTS

Across all trials, $F_{z1}$ averaged 2.13 BW ± 0.31 sd, with the highest values observed in the control condition. Of the three cushioned conditions, $F_{z1}$ was highest in the CS condition (Table 1b, Figure 2). Mean peak values of $F_{z1}$ were not significantly correlated with in vitro impact test results. ANOVA found significant differences among shoe conditions (p<0.01) but not among trials. Post-hoc analysis revealed that $F_{z1}$ was significantly lower in CM than in C0. CS and CH were not significantly different from C0, nor from one another. Furthermore, $F_{z1}$ occurred later in the stance phase under more compliant cushioning conditions. Timing differences among the experimental conditions were statistically significant (p<0.05), except for that between CS and CM.
On average, the two component model accounted for 98% of the intra-step variance in $F_z$. Peaks in $F_{HF}$ were synchronous with $F_{Z1}$ in all but C0, where there was a statistically significant delay of 3 ms (p<0.01). Mean peak values of $F_{HF}$ for each condition were smaller in magnitude than $F_{Z1}$ and were significantly correlated (p<0.05) with impact test Peak G scores and heel cushioning stiffness (Table 1c; Figure 2). $F_{HF}$ was significantly higher (p<0.5) in C0 than in all the cushioned shoes and was also significantly higher (p<0.05) in CH than in CS.

<table>
<thead>
<tr>
<th>Condition</th>
<th>(a) Properties</th>
<th>(b) $F_{Z1}$ †</th>
<th>(c) $F_{HF}$ †</th>
</tr>
</thead>
<tbody>
<tr>
<td>CS “Soft”</td>
<td>Peak G* 9.6 65</td>
<td>2.14 ± 0.16 14.9 ± 0.8</td>
<td>1.03 ± 0.13 13.9 ± 0.6</td>
</tr>
<tr>
<td>CM “Medium”</td>
<td>Stiffness kNm⁻¹ 12.3 97</td>
<td>2.05 ± 0.15 14.3 ± 0.9</td>
<td>1.09 ± 0.14 13.1 ± 0.7</td>
</tr>
<tr>
<td>CH “Hard”</td>
<td>15.1 152</td>
<td>2.05 ± 0.17 13.3 ± 0.7</td>
<td>1.11 ± 0.16 12.0 ± 0.6</td>
</tr>
<tr>
<td>C0 Control</td>
<td>25.0 439</td>
<td>2.28 ± 0.14 8.7 ± 0.7</td>
<td>1.65 ± 0.14 9.0 ± 0.5</td>
</tr>
</tbody>
</table>

* ASTM F1976-99 Impact Test † mean ± within-condition standard deviation

**DISCUSSION**

The findings of this study are consistent with previous reports showing that the $F_{Z1}$ peak occurs later in the stance phase in more cushioned shoes but that its magnitude differs little among different shoe conditions (Clarke et al., 1983; Nigg and Bahlsen, 1988; Snel et al., 1985). The trend for uncushioned shoes and those with very soft cushioning to produce higher $F_{Z1}$ scores than shoes with intermediate cushioning stiffnesses has also been previously reported (e.g. Nigg et al., 1987). The high frequency component $F_{HF}$ was coincident with $F_{Z1}$ but averaged 57% of $F_{Z1}$’s magnitude. Unlike $F_{Z1}$ but consistent with **in vitro** impact tests and the predictions of contact theory, $F_{HF}$ increased with increasing cushioning stiffness.

The two-component model of $F_z$ does not account for the inertial contributions of body segments, non-linear cushioning stiffnesses or compliance in the underlying surface. Nor does it remove the fundamental limitation of the force-plate as a cushioning evaluation tool – the ground reaction force reflects the average acceleration of the whole body, and is not specific to the lower extremity. These limitations notwithstanding, the model does serve to demonstrate that shoe cushioning affects measurable features of the ground reaction force during running and suggests an explanation for the counterintuitive effects of cushioning on $F_{Z1}$.

That explanation requires that $F_{Z1}$ is a composite of both heel impact and lower frequency force components. Assuming that shoes behave the same way **in vivo** as they do **in vitro**, softer cushioning both reduces and delays the impact peak. **In vivo**, the impact peak is summed with the low-frequency force pulse. The nonlinear summation depends on the cushioning stiffness. It tends to exaggerate the $F_{Z1}$ impact peak of softer shoes because a later impact peak is summed with a higher value of the low frequency force component.

Acceptance of the suggestion that $F_{Z1}$ is not an appropriate measure of impact magnitude during running has numerous implications, not least of which is that the conclusions of previous studies using $F_{Z1}$ as a measure of cushioning effects are probably not reliable. It also implies that the effective mass involved in heel impact is less than previously calculated and that **in vitro** impact tests based on that calculation are less valid than once thought. The hypothesis of active regulation of impact forces through kinematic adaptations would also need to be revisited.

**REFERENCES**